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13. ABSTRACT *(Maximum 200 words)*
To determine the effects of load carriage and walking speed on stride parameters and the coordination of trunk movements, twelve subjects walked on a level treadmill at a range of walking speeds (0.6 m·s-1 - 1.6 m·s-1) with and without a backpack containing 40% of their body mass. It was hypothesized that compared to unloaded walking load carriage decreases transverse pelvic and thoracic rotation, the mean relative phase between pelvic and thoracic rotations, and increases hip excursion. In addition, it was hypothesized that these changes would coincide with a decreased stride length and increase stride frequency. The findings supported the hypotheses. It was additionally hypothesized that the increased MOI of the upper body caused by the added mass of the backpack would result in an increase in upper body torque, an increase in lower body torque, and an increase in net body torque. Higher levels of upper body torque were observed in the backpack condition compared to the no backpack condition. However, the increase in upper body torque was 225%, while upper body MOI increased by 400%. In contrast to our hypothesis, decreased lower body torque was also observed in the backpack condition compared to the no backpack condition Dimensionless analyses indicated that there was a significantly larger contribution of hip excursion and smaller contribution on transverse plane pelvic rotation to increases in stride length during load carriage. In addition, there was a significant effect of load carriage on the amplitudes of transverse pelvic and thoracic rotation and the relative phase of pelvic and thoracic rotation. It was concluded that the shorter stride length and higher stride frequency observed when carrying a backpack is the result of decreased pelvic rotation. The shorter stride length and higher stride frequency observed during load carriage are associated with changes in transverse plane kinematics. These changes in transverse plane kinematics (decreased pelvic and thoracic rotation) may emerge as a consequence of the dynamics required to minimize torque production in the upper body which, in turn, may reduce the amount of muscle force required to control the load and the potential for injury.

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THE EFFECT OF WALKING SPEED AND ADDING A BACKPACK ON TRUNK DYNAMICS DURING TREADMILL WALKING

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BACKGROUND

Studies on the effects of load carriage on gait have shown changes in stride parameters that include increased stride frequency, double support time, and decreased stride length (7, 9, 12). Load carriage also results in increases in metabolic cost (12), lower limb joint reaction force, ground reaction force (4, 16), and an increased likelihood of low back injuries and stress fractures (8). Little is known, however, about the underlying mechanisms by which load influences gait parameters, and the associated changes in metabolic cost, reactive forces, and injury potential. This is particularly true for the changes in trunk dynamics and coordination of the upper and lower body--variables on which load may have immediate and profound effects. An understanding of the adaptations in dynamics and coordination patterns required for load carriage should provide more information on the relationships between kinetic, kinematic, and physiological variables.

In only a few experiments have researchers attempted to understand the mechanisms by which carrying a backpack influences gait patterns. The immediate goal of our research is to gain an understanding of these mechanisms. Specifically, the purpose of the present study is to investigate the underlying mechanisms by which carrying a load influences stride parameters.

ACKNOWLEDGMENTS

The authors would like to thank Masayoshi Kubo for his assistance in data collection and Suzanne Lynch for her assistance in preparing this technical report.

LIST OF SYMBOLS

Symbol	Name
u	Walking speed ($m \cdot s^{-1}$)
a	Stride length (m)
s	Stride frequency (Hz)
h	Leg length (m)
g	Acceleration due to gravity ($m \cdot s^{-1}$)
ϕ_P	Pelvic rotation (Deg)
ϕ_L	Hip excursion (Deg)
ϕ_T	Thoracic rotation (Deg)
ϕ_{TR}	Trunk rotation (Deg)
θ_s	Angular displacement ($rad \cdot s^{-1}$)
ω_s	Angular velocity ($rad \cdot s^{-1}$)
$\ddot{\phi}_P$	Pelvic angular acceleration ($rad \cdot s^{-1}$)
$\ddot{\phi}_T$	Thoracic angular acceleration ($rad \cdot s^{-1}$)
P_a	Phase angle (Deg)
Φ_{cont}	Continuous relative phase (Deg)
Φ_{disc}	Discrete relative phase (Deg)
$\hat{u}$	Dimensionless walking speed
$\hat{a}$	Dimensionless stride length
$\hat{s}$	Dimensionless stride frequency
DPR	Dimensionless pelvic rotation
$DThR$	Dimensionless thoracic rotation
$T_{LBnetCCW}$	Net lower body counterclockwise torque ($N \cdot m$)
$T_{LBPeakCCW}$	Peak lower body counterclockwise torque ($N \cdot m$)
T_{Net}	Net body torque ($N \cdot m$)
$T_{UBnetCCW}$	Net upper body counterclockwise torque ($N \cdot m$)
$T_{UBPeakCCW}$	Peak upper body counterclockwise torque ($N \cdot m$)

EXECUTIVE SUMMARY

To determine the effects of load carriage and walking speed on stride parameters and the coordination of trunk movements, 12 subjects walked on a treadmill at a range of walking speeds ($0.6\ m \cdot s^{-1}$ - $1.6\ m \cdot s^{-1}$) with and without a backpack containing 40% of their body mass. It was hypothesized that compared to unloaded walking, load carriage decreases transverse pelvic and thoracic rotation, the mean relative phase between pelvic and thoracic rotations, and increases hip excursion. In addition, it was hypothesized that these changes would coincide with a decreased stride length and increased stride frequency. The findings supported the hypotheses. Dimensionless analyses indicated that there was a significantly larger contribution of hip excursion and smaller contribution of transverse plane pelvic rotation to increases in stride length during load carriage. In addition, there was a significant effect of load carriage on the amplitudes of transverse pelvic and thoracic rotation and the relative phase of pelvic and thoracic rotation. It was additionally hypothesized that the increased MOI of the upper body caused by the added mass of the backpack would result in an increase in upper body torque, an increase in lower body torque, and an increase in net body torque. Higher levels of upper body torque were observed in the backpack condition compared to the no backpack condition. However, upper body torque in the backpack condition was not as high as estimates based solely on the increase in upper body MOI caused by adding the backpack. The addition of the backpack resulted in an average increase in upper body torque of about 225%, while upper body MOI increased by over 400%. In contrast to our hypothesis, decreased lower body torque was also observed in the backpack condition compared to the no backpack condition.

Previous literature suggests that during unloaded walking, the upper and lower body counter-rotate to reduce the net angular momentum of the body. The more in-phase pattern between transverse plane pelvic and thoracic rotation observed during load carriage compared to unloaded walking indicates this may not be the case. In contrast, the differences between upper body torque and lower body torque in loaded and unloaded walking suggest a result of loaded walking is to minimize upper body torque in the transverse plane, rather than to counterbalance upper body torque (as is the case in unloaded walking).

It was concluded that the shorter stride length and higher stride frequency observed when carrying a backpack is the result of decreased pelvic rotation. During unloaded walking, increases in pelvic rotation contribute to increases in stride length with increasing walking speed. The decreased pelvic rotation during load carriage requires an increased hip excursion to compensate. However, the increase in hip excursion is insufficient to fully compensate for the observed decrease in pelvis rotation, requiring an increase in stride frequency during load carriage to maintain a constant walking speed. The decrease in pelvic rotation is likely the result of adaptations in the gait that minimize lower body torque, in an effort to minimize the amount of torque potentially transmitted from the lower body to the upper body.

The present study demonstrates that the shorter stride length and higher stride frequency observed during load carriage are associated with changes in transverse plane kinematics. In addition, these changes in transverse plane kinematics (decreased pelvic and thoracic rotation) may emerge as a consequence of the dynamics required to minimize torque production in the upper body which, in turn, may reduce the amount of muscle force required to control the load and the potential for injury.

INTRODUCTION

Studies on the effects of load carriage on gait have shown changes in stride parameters that include increased stride frequency, double support time, and decreased stride length (7, 9, 12). Load carriage also results in increases in metabolic cost (12), lower limb joint reaction force, ground reaction force (4, 16), and an increased likelihood of low back injuries and stress fractures (8). Little is known, however, about the underlying mechanisms by which load influences gait parameters, and the associated changes in metabolic cost, reactive forces, and injury potential. This is particularly true for the changes in trunk dynamics and coordination of the upper and lower body--variables on which load may have immediate and profound effects. An understanding of the adaptations in dynamics and coordination patterns required for load carriage should provide more information on the relationships between kinetic, kinematic, and physiological variables.

In only a few experiments have researchers attempted to understand the mechanisms by which carrying a backpack influences gait patterns. The immediate goal of our research is to gain an understanding of these mechanisms. Researchers have reported that carrying loads greater than 50% body weight while walking at high speeds (about 1.78 $m \cdot s^{-1}$) results in a shorter stride length than when not carrying a load at similar speeds (9,12). No significant differences in stride length have been reported at slower walking speeds (about 1.35 $m \cdot s^{-1}$; 5, 11). The reason for this differential effect of load and walking speed on stride length is unclear. One purpose of the present study is to investigate the mechanisms by which carrying a load influences stride parameters.

Wagenaar and Beek (19) demonstrated that increases in stride length with increasing walking speed are associated with changes in transverse plane pelvic and thoracic rotation. In specific, increasing walking speed greater than approximately 1.0 $m \cdot s^{-1}$ (walking speed was increased from 0.25 $m \cdot s^{-1}$ to 1.5 $m \cdot s^{-1}$) resulted in an increase in transverse pelvic rotation, whereas no significant changes in transverse thoracic rotation were found. Dimensionless pelvic rotation (pelvic rotation divided by hip excursion) decreased when walking speed was increased from 0.25 to 0.75 $m \cdot s^{-1}$ and remained constant when walking speed increased greater than 1.0 $m \cdot s^{-1}$. These findings indicated that in addition to an increase in hip excursion, the increase in stride length at walking speeds greater than 1.0 $m \cdot s^{-1}$ requires a larger contribution of transverse pelvic rotation.

Based on the results of Wagenaar and Beek (19), the shorter stride length during load carriage at higher walking speeds (9) suggests a decrease in transverse pelvic rotation. Our first hypothesis was that carrying a backpack results in less transverse pelvic rotation than unloaded walking.

To maintain a constant speed with less pelvic rotation, the individual can either increase hip excursion in the sagittal plane and, hence, maintain stride length, or increase stride frequency to maintain walking speed (6). Our second hypothesis was

that one or both of these mechanisms will occur to maintain walking speed when carrying a backpack.

In addition, Wagenaar and Beek (19) showed that increasing walking speed from 0.25 $m \cdot s^{-1}$ to 1.5 $m \cdot s^{-1}$ was associated with a gradual transition from a more in-phase relationship of pelvic and thoracic rotation (rotating in the same direction at the same time) to a more out-of-phase (rotating in opposite directions) relationship at approximately 0.8 $m \cdot s^{-1}$. This transition was accompanied by a decrease in the standard deviation (stability) of relative phase at intermediate walking speeds, suggesting the existence of two distinct coordination patterns during walking (17). It was suggested that the upper torso (head, arms and thorax) and lower torso (legs and pelvis) counter-rotate at higher walking speeds in order to reduce the net angular momentum of the body (14).

A major kinetic influence of carrying a backpack is to increase the transverse plane MOI of the upper body and, because torque is the product of MOI and angular acceleration, to potentially increase the amount of torque produced by the upper body. Balancing torques would be necessary to minimize the net angular momentum of the body. Therefore, load carriage presents an opportunity to investigate the relationship between torque production, angular momentum, and emergent coordination patterns.

Because adding a backpack will increase the MOI of the thorax, less thoracic rotation may be needed to counterbalance pelvic rotation. Combined with the anticipated reduction in transverse pelvic rotation and concomitant decrease expected in pelvic angular momentum, our third hypothesis was that there will be less thoracic rotation, trunk rotation and decreased counter-rotation of the pelvis and thorax at the higher walking speed when carrying a backpack.

Townsend (15) presented a model of unloaded human locomotion that predicted an increase in the MOI of the upper and lower body with increasing walking speed. The prediction was based on larger amplitudes of arm and leg swing resulting in a greater distance between the center of mass of the limbs and the longitudinal axis of the body. In addition, if there is an increase in stride frequency and stride length associated with greater walking speed during load carriage, greater accelerations of the trunk segments would also be expected. Our fourth hypothesis was that increasing walking speed would result in an increase in upper and lower body torque during loaded and unloaded walking.

Our fifth hypothesis was that carrying a backpack would result in an increase in upper body torque due to the increased MOI associated with the load. In addition, if there is a more in-phase pattern of pelvic and thoracic rotation during load carriage, the upper and body may be more rigidly linked. If the upper and lower bodies are rigidly linked, an increase in lower body torque would also be expected. Therefore, it was hypothesized that load carriage would result in an increase in lower body torque. In addition, the in-phase torques generated between the upper and lower body should result in a net body torque that is increased during load carriage.

METHODS

RESEARCH VOLUNTEERS

Fourteen healthy subjects participated in this study. Two subjects were excluded from the final analysis due to technical problems during data collection. Five males and seven females (mean age 26 years 7.1 SEM) were used in the final analysis. Subjects were selected from the Boston University community, participated in strenuous physical exercise at least three times per week, and had no orthopedic disorders or complicating medical histories. Prior to participation, subjects gave informed consent in accordance with the policies of Boston University Institutional Review Board. The research was conducted in adherence with the provisions of 45 CFR Part 46.

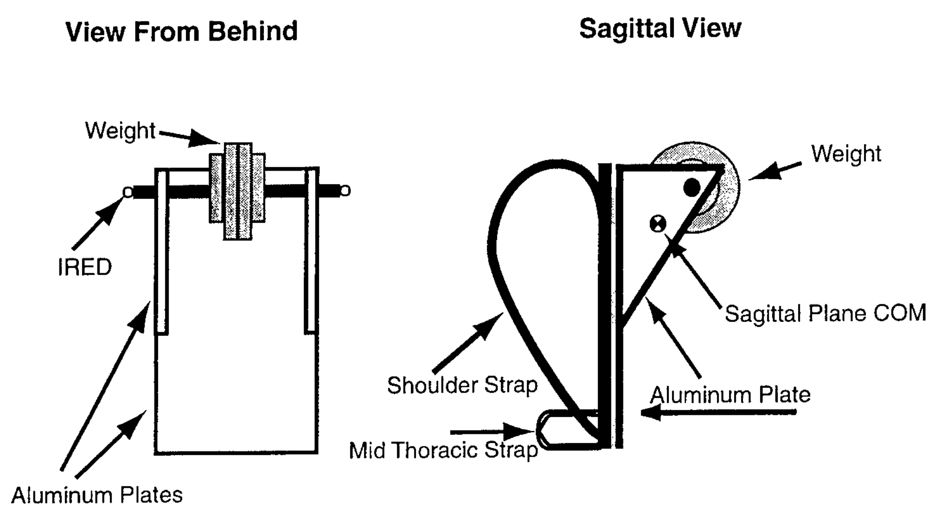
DATA COLLECTION

Data were collected in the Barreca Motion Laboratory at Boston University. Prior to data collection, anthropometric measures of total leg, shank, thigh and arm length, as well as hip and shoulder width were taken. Body mass and height were measured using a balance scale.

EXPERIMENTAL BACKPACK

The backpack frame was constructed of rigid plastic and designed to make contact only with the thorax (Figure 1). An aluminum rod was attached to the frame to hold the weight at shoulder height as close to the subject as possible. Two shoulder straps and a mid-thoracic strap minimized pack movement in relation to the thorax. The total weight of the backpack was adjusted to 40% of the subject's body weight. The weight was chosen to fall in the range of weights normally tested in backpack experiments (2, 16).

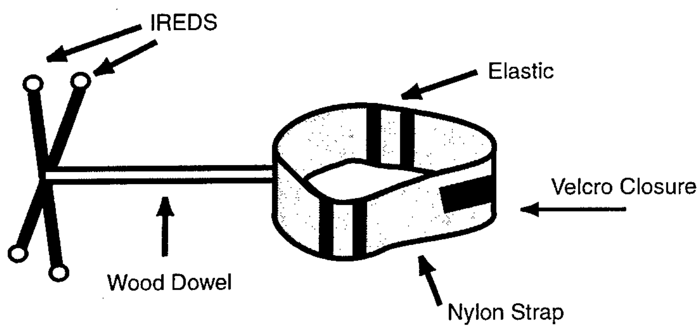
Figure 1. Illustration of backpack designed to make contact solely with the thorax



EXPERIMENTAL SETUP

Infrared Light Emitting Diodes (IREDS) were placed bilaterally on the subject's zygomatic processes, acromion processes, mid thighs, lateral femoral condyles, lateral malleoli, and ulnar styloid processes. Transverse pelvic and thoracic rotations were recorded using two custom made T-squares (18). The top of the strap of the T-Square that tracked thoracic movement was approximately ½ inch below the subject's armpit. The top of the T-Square that tracked pelvic motion was aligned at the level of anterior superior iliac spine. A validity test showed that, on average, the motion of the T-Squares tracked the motion of the pelvis and thorax to within 0.02°. IREDS were placed on each end of the T-Squares (Figure 2), bilaterally on the center of mass of the backpack in the sagittal plane, and on the ends of the bar holding the weight.

Figure 2. Illustration of T square designed to track the movement of the pelvis; a similar T square was used to track the motion of the thorax



DATA COLLECTION

Three-dimensional kinematic data were collected at 100 Hz through an Optotrak 3020 System (Northern Digital, Waterloo, Ont., Canada). Two position sensors were placed on each side of the treadmill, approximately 3 meters from the treadmill. Subjects walked on an instrumented Kistler/Trotter treadmill (Gaitway model; Kistler Instrument Corporation, Amherst, USA) capable of measuring vertical ground reaction force (VGRF). On average, the belt speed varied 0.01 m·s-1 (<2.0%) within a speed condition regardless of load.

A Polar Heart Rate Monitor was used to assess if heart rate exceeded 80% of the age predicted maximum heart rate, indicating overexertion (9). The experiment was to be terminated in this event. The monitor is composed of an electrode that is placed around the chest of the subject and a wristwatch that reports the subject's heart rate.

Subjects were asked to walk with and without a backpack at six different speeds. The sequence of backpack or no backpack condition was balanced across subjects. There were a total of six speed conditions for each of the two backpack conditions. The speeds were systematically increased 0.6 $m \cdot s^{-1}$ to 1.6 $m \cdot s^{-1}$ in .2 $m \cdot s^{-1}$ increments and then decreased in order to investigate hysteresis effects, resulting in a 6x2x2 design (6 speed conditions, 2 backpack conditions, 2 direction of walking speed changes: increasing and decreasing). Subjects walked at each speed for approximately 3 minutes. During the last 30 seconds, kinematic and kinetic data were collected.

DATA PROCESSING

Heel strike was determined to be the first frame of the VGRF time series that was greater than 7% of the peak VGRF; toe off was the last frame that the VGRF was greater than 7% of the peak VGRF (1). Heel strike was used to mark the beginning and end of each stride. Each stride of data was time-normalized to percentage of stride. Heel strike and toe off data were used to calculate stride length and stride frequency.

The raw kinematic data were converted into three-dimensional data by means of the Optotrak system software. Missing data were interpolated using a cubic spline. If there were more than 15 consecutive frames of missing data within any particular stride, that stride was discarded because the interpolation was not reliable. The interpolation procedure was validated against known values before its use and demonstrated a maximum error of 1.0 mm. Aberrant force-plate data would occur if both feet were on the same force-plate at the same time. There were a total of 4435 strides of data collected, of which 3.0% were eliminated due to missing kinematic or aberrant force-plate data. After interpolation, the data were filtered at 5 Hz (low pass second order Butterworth).

Sagittal plane hip rotation and transverse plane pelvic and thoracic rotation were calculated from the filtered and interpolated time series (Figures 3 and 4). Peak pelvic and thoracic angular accelerations were calculated as the second derivative of pelvic and thoracic rotation. The transverse plane axis of rotation of the pelvis was assumed to be located at the spine, midway between the greater trochanters of the left and right hips. The transverse plane axis of rotation of the thorax was also assumed to be located in the spine, midway between the shoulders.

Hip angle was calculated as the angle between the thigh and vertical. Hip excursion is the difference between peak hip flexion and peak hip extension. Trunk rotation is the difference between transverse pelvic rotation and thoracic rotation. Angular velocity was calculated from the angular displacement data. The variability of the angle between the backpack and thorax is a measure of backpack movement in relation to the thorax. On average the angle varied by less than 0.34° within a trial.

Continuous relative phase was calculated using the method described in van Emmerik and Wagenaar (18). Both angular position and velocity data were time-normalized to percentage of stride and then normalized to minimal and maximal

Figure 3. Example time series of pelvic and thoracic rotation in the no backpack condition

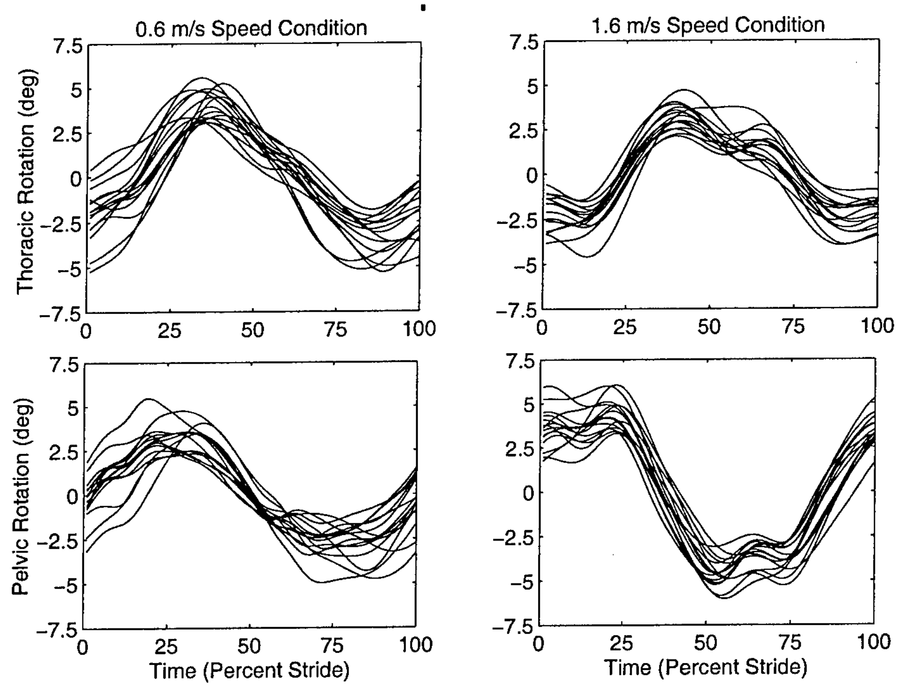
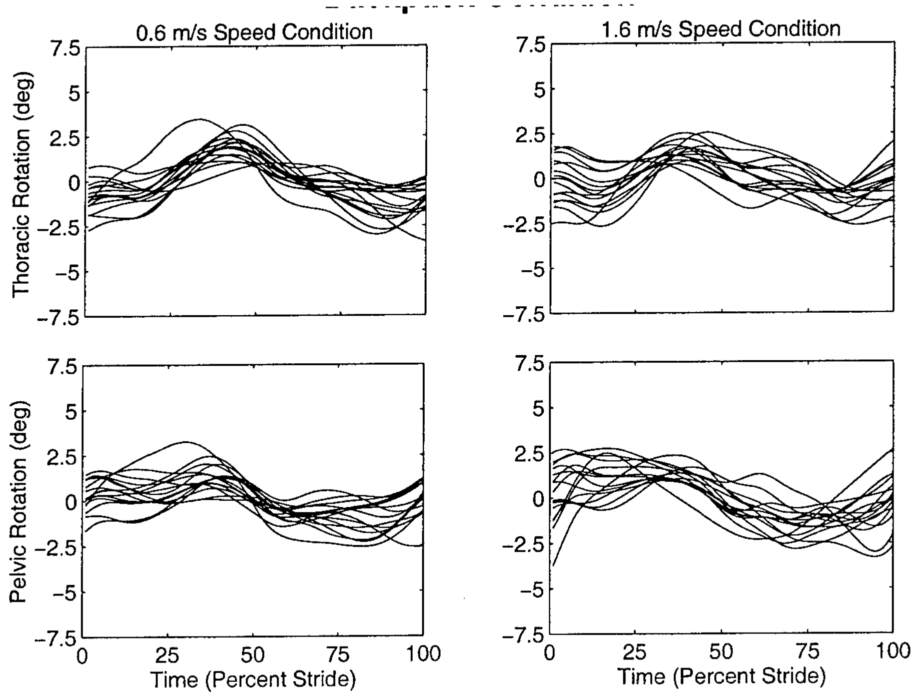


Figure 4. Example time series of pelvic and thoracic rotation in the backpack condition

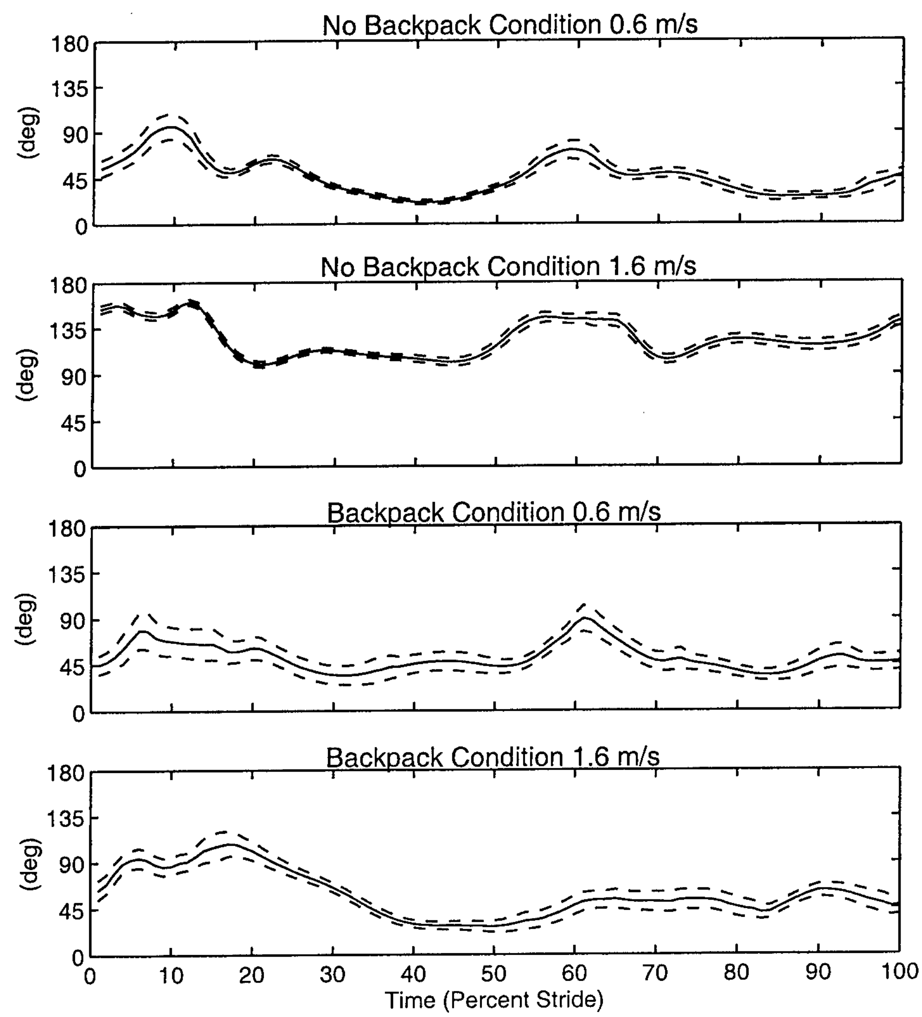


amplitudes (–1 and 1) within each stride. After normalization, the phase angles for the pelvis and the thorax for every point in time were calculated using Equation 1:

$$P_a = \arctan(\omega_s(t)/\theta_s(t)) \qquad (1),$$

where P_a represents phase angle, t time, ω_s angular velocity, and θ_s angular displacement. Continuous relative phase between transverse pelvic and thoracic rotation is the difference between the pelvic and thoracic phase angles, and is calculated in the range of 0º-180°, and averaged across each stride (Figure 5). Discrete relative phase between pelvic and thoracic rotation at heel strike was determined using Equation 1, and substituting heel strike as t.

Figure 5. Example relative phase time series for fast and slow walking in the backpack and no backpack conditions



DIMENSIONLESS ANALYSIS

Dimensionless walking speed ($\hat{u} = u(g * h)^{-.5}$), stride length ($\hat{a} = (a/h)^{.5}$), and stride frequency ($\hat{s} = a(h/g)^{.5}$) were calculated to allow for normalization with respect to leg length (h) and walking velocity (u), where a represents stride length, s, stride frequency, and g, the acceleration due to gravity. Dimensionless pelvic rotation ($DPR = \phi_P / \phi_L$) provided an estimate of the relative contribution of the transverse pelvic rotation and hip excursion to the stride, and dimensionless thoracic rotation ($DThR = \phi_T / \phi_P$) of the difference in amplitude between transverse thoracic and pelvic rotation (ϕ_P and ϕ_T represent the amplitude of pelvic and thoracic rotation, and ϕ_L represents hip excursion). To estimate the invariance in timing of the transverse pelvic and thoracic rotations, dimensionless trunk rotation was calculated using Equation 2:

$$DTR = \arccos[(\phi_P^2 + \phi_T^2 + \phi_{TR}^2) / 2\phi_P\phi_L] / \Phi \qquad (2),$$

ϕ_{TR} represents maximal trunk angle, and Φ the relative phase between pelvic and thoracic rotation (for derivation see Wagenaar and Beek, 19).

TORQUE CALCULATION

The MOIs of the upper and lower body were calculated using Equation 3:

$$I = \sum_{i=1}^{n} m_i r_i^2 \qquad (3)$$

m represents the mass of the segment, r, the distance between the center of mass of that segment and the axis of rotation, and n, the number of segments. Segment mass and the position of each segment's center of mass were based on anthropometrics and calculated from estimates given by Dempster (3). The lower body included five segments: two shanks, two thighs, and the pelvis. The feet and shank were considered one segment. The upper body included three segments: two arms and the thorax. The upper and lower arm were considered to be one segment. The mass of the head was included in the mass of the thorax segment. The contribution of the backpack to the MOI of the upper body was calculated using the parallel axis theorem.

On average, the upper body rotates the same distance clockwise per stride as it rotates counterclockwise; consequently, the net torque of the upper body per stride was expected to be zero. In order to obtain a measure of net torque of the upper body per stride that was not zero, the net torque of the upper body was calculated in the

clockwise and counterclockwise directions ($T_{UBnetCW}$ & $T_{UBnetCCW}$) separately. Torque was calculated for each percentage of stride as the product of MOI and angular acceleration ($T = I\alpha$). $T_{UBnetCCW}$ was calculated as the sum of all the positive upper body torque per stride. Similarly, $T_{UBnetCW}$ is the sum of negative torque, also calculated per stride. Upper body peak torque for each stride was also determined in both the clockwise and counterclockwise directions ($T_{UBPeakCW}$ & $T_{UBPeakCCW}$). Net and peak lower body torque in both the clockwise and counterclockwise directions ($T_{LBnetCW}$, $T_{LBnetCCW}$, $T_{LBPeakCW}$ & $T_{LBPeakCCW}$) were calculated in the same manner as the upper body torque variables.

In addition to calculating net and peak upper and lower body torque, we also determined the net torque of the body. The net torque of the body (T_{Net}) represents the sum of all torque acting on the trunk per stride. Upper and lower body torque was added for each percentage of stride (T_{Neti}), and then the sum of the absolute value of T_{Neti} for each stride was determined (Equation 4).

$$T_{net} = \sum_{i=1}^{100} \left| T_{LB_i} + T_{UB_i} \right| \quad (4)$$

i represents the percentage of stride, T_{UB_i} and T_{LB_i} the torque of the upper and lower body for frame i.

Predictions of upper body torque (T_{PRED}) were calculated for each backpack MOI condition as the product of MOI of the upper body (including the contribution of the backpack) and the thoracic angular acceleration in the no backpack condition (Equation 5):

$$T_{PRED} = I_{UP}\ddot{\theta}_{ThorNBP} \quad (5)$$

I_{UP} represents the MOI of the upper body, and $\ddot{\theta}_{ThorNBP}$, thoracic angular acceleration in the no backpack condition.

STATISTICS

A 6x2x2 repeated measures Analysis of Variance (ANOVA) with three within-subject effects was used to test for the main effects of speed (6 levels), backpack (2 levels) and hysteresis (2 levels) on the dependent variables, and for speed x backpack interaction effects. Alpha was set to 0.05 for main and interaction effects. If there were no statistically significant effect of hysteresis, the data from the increasing and

decreasing walking speed parts of the experiment were pooled for the rest of the analysis. If a significant speed x backpack interaction effect was found, a Duncan post-hoc was used to determine which backpack and speed conditions were statistically different. In addition, paired t-tests were used to determine the walking speeds at which there were significant differences between the backpack and no backpack conditions. However, if there was a statistically significant effect of hysteresis, the data for the same speed condition in the increasing and decreasing portions of the experiment were analyzed separately in the post hoc analysis. Based on the number of comparisons in the post hoc analysis, alpha was set to 0.0045 (13).

RESULTS

Table 1. P values for the main effects of backpack, walking speed and for the speed * backpack interaction for each dependent variable.

Variable	Main Effect of Backpack p	Main Effect of Speed p	Interaction of Speed * Backpack p
Stride Length (m)	0.0001	0.0001	0.6941
Dimensionless Stride Length	0.0001	0.0001	0.6564
Stride Frequency (Hz)	0.0001	0.0001	0.8186
Dimensionless Stride Frequency	0.0001	0.0001	0.8345
Pelvic Rotation (º)	0.0001	0.0001	0.0337
Hip Excursion (º)	0.0001	0.0001	0.0202
Dimensionless Pelvic Rotation	0.0001	0.0001	0.0152
Thoracic Rotation (º)	0.0001	0.0003	0.0224
Dimensionless Thoracic Rotation	0.0038	0.0001	0.0369
Trunk Rotation (º)	0.0001	0.0001	0.0001
Continuous Relative Phase (º)	0.0001	0.0001	0.0001
Discrete Relative Phase (º)	0.0001	0.0001	0.0001
Counterclockwise Pelvic Rotation ($rad\cdot s^{-1}$)	0.0015	0.0001	0.0001
Net Counterclockwise Lower Body Torque (N·m)	0.0001	0.0001	0.3300
Peak Counterclockwise Lower Body Torque (N·m)	0.0013	0.0001	0.8365
Counterclockwise Thoracic Rotation ($rad\cdot s^{-1}$)	0.0001	0.0001	0.0016
Net Counterclockwise Upper Body Torque (N·m)	0.0001	0.0001	0.0001
Peak Counterclockwise Upper Body Torque (N·m)	0.0001	0.0001	0.0001
Net Body Torque (N·m)	0.0002	0.0001	0.0001

Carrying a backpack resulted in a shorter stride length than walking without a backpack (Table 2). In contrast, a longer stride length was observed at higher walking speeds than at lower walking speeds in both the backpack and no backpack conditions. A significant effect of hysteresis indicates that stride length was shorter at the same walking speed during the decreasing walking speed portion of the experiment than during the increasing walking speed portion. The purpose of this experiment was to determine the underlying mechanisms that influence stride length during load carriage.

Table 2. Stride length, a (m): mean (standard error).

	INCREASING SPEED						DECREASING SPEED				
	0.6 $m \cdot s^{-1}$	**0.8 $m \cdot s^{-1}$**	**1.0 $m \cdot s^{-1}$**	**1.2 $m \cdot s^{-1}$**	**1.4 $m \cdot s^{-1}$**	**1.6 $m \cdot s^{-1}$**	**1.4 $m \cdot s^{-1}$**	**1.2 $m \cdot s^{-1}$**	**1.0 $m \cdot s^{-1}$**	**0.8 $m \cdot s^{-1}$**	**0.6 $m \cdot s^{-1}$**
Backpack	0.85 (0.01) J	1.02 (0.01) H	1.16 (0.01) F	1.27 (0.01) D	1.37 (0.01) B	1.48 (0.01) A	1.37 (0.01) B	1.28 (0.01) C	1.18 (0.01) E	1.05 (0.01) G	0.90 (0.01) I
No Backpack	0.89 (0.01) J	1.03 (0.01) H	1.19 (0.00) F	1.30 (0.00) E	1.42 (0.00) B	1.55 (0.00) A	1.41 (0.00) C	1.31 (0.00) D	1.19 (0.00) F	1.05 (0.01) G	0.91 (0.01) I

Main Effect of Backpack: <0.0001
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: 0.6941
Significant Effect of Hysteresis: 0.0009
Within a backpack condition, speeds with the same letter are not significantly different.

Dimensionless stride length is calculated by dividing stride length by leg length, and provides a measure of stride length that accounts for differences in anthropometry between subjects. Increasing walking speed was associated with an increase in dimensionless stride length (Table 3). In contrast, carrying a backpack resulted in a decrease in dimensionless stride length.

Table 3. Dimensionless stride length, $\hat{a}$: mean (standard error).

	0.6 m·s^{-1}	**0.8 m·s^{-1}**	**1.0 m·s^{-1}**	**1.2 m·s^{-1}**	**1.4 m·s^{-1}**	**1.6 m·s^{-1}**
Backpack	1.09 (0.007) F	1.28 (0.006) E	1.45 (0.006) D	1.59 (0.006) C	1.73 (0.007) B	1.83 (0.011) A
No Backpack	1.10 (0.001) F	1.28 (0.004) E	1.47 (0.004) D	1.61 (0.001) C	1.75 (0.001) B	1.91 (0.001) A

Main Effect of Backpack: <0.0001
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: 0.6564
Within a backpack condition, speeds with the same letter are not significantly different.

Consistent with previous research, increasing walking speed was associated with an increase in stride frequency (Table 4). The decrease in stride length observed in the backpack condition requires an increase in stride frequency to maintain a constant walking speed. As expected from the significant effect of hysteresis on stride length, there was a significant effect of hysteresis on stride frequency, indicating stride frequency was higher at the same walking speed during the decreasing walking speed portion of the experiment than during the increasing walking speed portion.

Table 4. Stride frequency, s (Hz): mean (standard error).

	INCREASING SPEED						DECREASING SPEED				
	0.6 $m \cdot s^{-1}$	**0.8 $m \cdot s^{-1}$**	**1.0 $m \cdot s^{-1}$**	**1.2 $m \cdot s^{-1}$**	**1.4 $m \cdot s^{-1}$**	**1.6 $m \cdot s^{-1}$**	**1.4 $m \cdot s^{-1}$**	**1.2 $m \cdot s^{-1}$**	**1.0 $m \cdot s^{-1}$**	**0.8 $m \cdot s^{-1}$**	**0.6 $m \cdot s^{-1}$**
Backpack	0.70 (0.01) H	0.79 (0.01) F	0.88 (0.01) D	0.94 (0.00) C	1.01 (0.00) B	1.08 (0.00) A	1.01 (0.00) B	0.94 (0.00) C	0.87 (0.00) E	0.77 (0.01) G	0.65 (0.01) I
No Backpack	0.68 (0.01) H	0.78 (0.01) F	0.86 (0.00) E	0.92 (0.00) C	0.98 (0.00) B	1.04 (0.00) A	0.98 (0.00) B	0.91 (0.00) D	0.86 (0.00) E	0.76 (0.00) G	0.64 (0.01) I

Main Effect of Backpack: <0.0001
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: 0.8186
Significant Effect of Hysteresis: 0.0006
Within a backpack condition, speeds with the same letter are not significantly different.

Dimensionless stride frequency provides a measure of stride frequency that accounts for differences in anthropometry between subjects. Increasing walking speed and carrying a backpack were both associated with an increased dimensionless stride frequency (Table 5).

Table 5. Dimensionless stride frequency, $\hat{s}$: mean (standard error).

	0.6 m·s^{-1}	**0.8 m·s^{-1}**	**1.0 m·s^{-1}**	**1.2 m·s^{-1}**	**1.4 m·s^{-1}**	**1.6 m·s^{-1}**
Backpack	0.19 (0.001) F	0.22 (0.001) E	0.25 (0.001) D	0.27* (0.001) C	0.29 (0.001) B	0.31* (0.001) A
No Backpack	0.19 (0.001) F	0.22 (0.001) E	0.25 (0.001) D	0.26 (0.001) C	0.28 (0.001) B	0.30 (0.001) A

Main Effect of Backpack: < 0.0001
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: 0.8345
Within a backpack condition, speeds with the same letter are not significantly different.

Increasing walking speed resulted in an increase in transverse plane pelvic rotation in both the backpack and no backpack conditions (Table 6). However, transverse plane pelvic rotation in the backpack condition was approximately half of what was observed at the same walking speed in the no backpack condition. During unloaded walking at higher walking speeds, transverse plane pelvic rotation contributes to the lengthening of the stride. Consequently, the lack of increase in pelvic rotation in the backpack condition may contribute the lack of increase in stride length.

Table 6. Transverse plane pelvic rotation, ϕ_P(Deg): mean (standard error).

	0.6 m·s^{-1}	**0.8 m·s^{-1}**	**1.0 m·s^{-1}**	**1.2 m·s^{-1}**	**1.4 m·s^{-1}**	**1.6 m·s^{-1}**
Backpack	4.29* (0.12) B	4.20* (0.10) B	4.24* (0.11) B	4.19* (0.09) B	4.40* (0.08) B	5.59* (0.13) A
No Backpack	8.62 (0.25) C	7.69 (0.15) D E	7.50 (0.12) E	7.95 (0.11) D	9.02 (0.01) B	10.91 (0.17) A

Main Effect of Backpack: <0.0001
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: 0.0337
Within a backpack condition, speeds with the same letter are not significantly different.
*Signifies differences between the backpack and no backpack condition at a specific walking speed.

At higher walking speeds, stride length can be maintained by increasing transverse plane pelvic rotation or by increasing hip excursion (sagittal plane leg swing angle). The lack of increase in transverse plane pelvic rotation in the backpack condition results in a need for increased hip excursion to maintain stride length (Table 7). However, if the increase in hip excursion is not sufficient, a decrease in stride length will likely result.

Table 7.: Hip excursion, ϕ_L (Deg): mean (standard error).

	INCREASING SPEED						DECREASING SPEED				
	0.6 $m \cdot s^{-1}$	**0.8 $m \cdot s^{-1}$**	**1.0 $m \cdot s^{-1}$**	**1.2 $m \cdot s^{-1}$**	**1.4 $m \cdot s^{-1}$**	**1.6 $m \cdot s^{-1}$**	**1.4 $m \cdot s^{-1}$**	**1.2 $m \cdot s^{-1}$**	**1.0 $m \cdot s^{-1}$**	**0.8 $m \cdot s^{-1}$**	**0.6 $m \cdot s^{-1}$**
Backpack	39.48 (0.37) F*	41.03 (0.39) E*	45.61 (0.40) D*	46.47 (0.37) C*	50.19 (0.27) B*	52.94 (0.31) A*	50.31 (0.33) B*	47.10 (0.32) C*	45.15 (0.36) D*	41.36 (0.39) E*	39.07 (0.41) F*
No Backpack	34.22 (0.33) H	36.49 (0.28) F	40.26 (0.26) D	42.11 (0.21) C	44.55 (0.20) B	47.35 (0.19) A	44.46 (0.19) B	42.67 (0.25) C	40.01 (0.27) D	37.62 (0.59) E	34.67 (0.33) G

Main Effect of Backpack: <0.0001
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: 0.0202
Significant Effect of Hysteresis: 0.0325
Within a backpack condition, speeds with the same letter are not significantly different.
*Signifies differences between the backpack and no backpack condition at a specific walking speed.

Dimensionless pelvic rotation is pelvic rotation divided by hip excursion, and provides a measure of transverse plane pelvic rotation relative to the sagittal plane leg swing angle. The decrease in dimensionless pelvic rotation observed in the backpack condition indicates the swinging of the leg (in the sagittal plane) contributes more to increases in stride length in the backpack condition than in the no backpack condition (Table 8).

Table 8. Dimensionless pelvic rotation, DPR: mean (standard error).

	0.6 m·s^{-1}	**0.8 m·s^{-1}**	**1.0 m·s^{-1}**	**1.2 m·s^{-1}**	**1.4 m·s^{-1}**	**1.6 m·s^{-1}**
Backpack	0.11* (0.003) A	0.10* (0.003) A	0.10* (0.003) B	0.09* (0.002) B	0.09* (0.002) B	0.11* (0.003) A
No Backpack	0.25 (0.008) A	0.021 (0.004) C	0.19 (0.003) D	0.19 (0.003) D	0.21 (0.002) C	0.23 (0.004) B

Main Effect of Backpack: <0.0001
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: 0.0152
Within a backpack condition, speeds with the same letter are not significantly different.
*Signifies differences between the backpack and no backpack condition at a specific walking speed.

Increasing walking speed resulted in a decrease in transverse plane thoracic rotation (Table 9). As expected, transverse plane thoracic rotation was less in the backpack condition than in the no backpack condition.

Table 9. Thoracic rotation, ϕ_T (Deg): mean (standard error).

	0.6 m·s^{-1}	**0.8 m·s^{-1}**	**1.0 m·s^{-1}**	**1.2 m·s^{-1}**	**1.4 m·s^{-1}**	**1.6 m·s^{-1}**
Backpack	4.57* (0.14) A	4.56* (0.10) A	4.41* (0.10) A B	4.32* (0.08) A B	4.11* (0.08) B	3.65* (0.09) C
No Backpack	9.31 (0.27) B	9.71 (0.15) A	9.24 (0.11) B	8.58 (0.09) C	8.16 (0.08) D	8.04 (0.13) D

Main Effect of Backpack: <0.0001
Main Effect of Speed: 0.0003
Interaction of Speed * Backpack: 0.0224
Within a backpack condition, speeds with the same letter are not significantly different.
*Signifies differences between the backpack and no backpack condition at a specific walking speed.

Dimensionless thoracic rotation provides a measure of the difference in amplitude between transverse thoracic and pelvic rotation. Values greater than 1.0 indicate transverse plane thoracic rotation is greater than transverse plane pelvic rotation; values less than 1.0 indicate pelvic rotation is greater than thoracic. The lesser values of dimensionless thoracic rotation observed in the backpack condition indicate the amplitude of thoracic rotation relative to the amplitude of pelvic rotation is less in the backpack condition than in the no backpack condition (Table 10).

Table 10. Dimensionless thoracic rotation, $DThR$: mean (standard error).

	0.6 m·s^{-1}	**0.8 m·s^{-1}**	**1.0 m·s^{-1}**	**1.2 m·s^{-1}**	**1.4 m·s^{-1}**	**1.6 m·s^{-1}**
Backpack	1.11 (0.03) A B	1.20 (0.03) B	1.18 (0.03) A B	1.18 (0.03) A B	1.07 (0.03) B	0.73 (0.03) C
No Backpack	1.18 (0.03) B	1.37 (0.03) A	1.36 (0.02) A	1.22 (0.03) B	1.03 (0.02) C	0.87 (0.03) D

Main Effect of Backpack: 0.0038
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: 0.0369
Within a backpack condition, speeds with the same letter are not significantly different.
*Signifies differences between the backpack and no backpack condition at a specific walking speed.

Trunk rotation is the difference between peak thoracic rotation in one direction and peak transverse plane pelvic rotation in the other direction. Increasing walking speed resulted in an increase in trunk rotation (Table 11). As expected, trunk rotation was less in the backpack condition than in the no backpack condition.

Table 11. Trunk rotation, ϕ_{TR} (Deg): mean (standard error).

	0.6 m·s^{-1}	**0.8 m·s^{-1}**	**1.0 m·s^{-1}**	**1.2 m·s^{-1}**	**1.4 m·s^{-1}**	**1.6 m·s^{-1}**
Backpack	3.41* (0.12) F	4.08* (0.13) E	4.51* (0.12) D	4.99* (0.11) C	5.33* (0.09) B	5.90* (0.11) A
No Backpack	7.83 (0.16) F	9.74 (0.14) E	11.59 (0.14) D	12.70 (0.13) C	14.04 (0.10) B	16.19 (0.14) A

Main Effect of Backpack: <0.0001
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: <0.0001
Within a backpack condition, speeds with the same letter are not significantly different.
*Signifies differences between the backpack and no backpack condition at a specific walking speed.

In the present study, continuous and discrete relative phase are measures of the amount of counter-rotation between the pelvis and thorax during walking. At slower walking speeds during both unloaded and loaded walking, there is little counter-rotation between the pelvis and thorax (relative phase < 90°) (Tables 12 and 13). Increasing walking speed results in a gradual transition from an in-phase pattern of pelvic and thoracic rotation (little counter-rotation) to an out-of-phase pattern (counter-rotation; relative phase > 90°). In contrast, no transition is observed in the backpack condition.

Table 12. Continuous relative phase, Φ_{cont} (Deg): mean (standard error).

	0.6 m·s^{-1}	**0.8 m·s^{-1}**	**1.0 m·s^{-1}**	**1.2 m·s^{-1}**	**1.4 m·s^{-1}**	**1.6 m·s^{-1}**
Backpack	48.18 (2.10) E	54.50 (1.58) D	62.86* (1.45) C	73.70* (1.51) B	80.39* (1.52) A	78.78* (1.53) A
No Backpack	54.33 (1.58) F	72.50 (1.72) E	95.21 (1.51) D	110.18 (1.20) C	118.95 (0.91) B	122.06 (1.13) A

Main Effect of Backpack: <0.0001
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: <0.0001
Within a backpack condition, speeds with the same letter are not significantly different.
*Signifies differences between the backpack and no backpack condition at a specific walking speed.

Table 13. Discrete relative phase, Φ_{disc} (Deg): mean (standard error).

	0.6 m·s^{-1}	**0.8 m·s^{-1}**	**1.0 m·s^{-1}**	**1.2 m·s^{-1}**	**1.4 m·s^{-1}**	**1.6 m·s^{-1}**
Backpack	42.72 (3.18) C	48.27 (2.42) C	58.53* (2.64) B	60.59* (2.31) B	66.31* (2.45) A B	71.54* (3.42) A
No Backpack	58.31 (2.86) E	78.66 (2.87) D	97.67 (2.45) C	118.89 (1.94) B	130.76 (1.56) A	133.85 (2.02) A

Main Effect of Backpack: <0.0001
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: <0.0001
Within a backpack condition, speeds with the same letter are not significantly different.
*Signifies differences between the backpack and no backpack condition at a specific walking speed.

As expected from the decrease in pelvic rotation and the increase in stride frequency observed in the backpack condition, counterclockwise pelvic acceleration was less in the backpack condition than in the no backpack condition (Table 14). Results for clockwise acceleration were equivalent to those for counterclockwise, but are omitted to prevent repetition.

Table 14. Counterclockwise pelvic angular acceleration, $\ddot{\phi}_P$ (rad·s^{-1}): mean (standard error).

	0.6 m·s^{-1}	**0.8 m·s^{-1}**	**1.0 m·s^{-1}**	**1.2 m·s^{-1}**	**1.4 m·s^{-1}**	**1.6 m·s^{-1}**
Backpack	3.26* (0.12) F	4.29* (0.12) E	5.14* (0.16) D	5.80* (0.13) C	6.55* (0.11) B	8.09 (0.17) A
No Backpack	5.85 (0.12) E	7.00 (0.12) D	7.68 (0.11) C	7.89 (0.10) C	8.40 (0.10) B	9.97 (0.15) A

Main Effect of Backpack: 0.0015
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: 0.8377
Within a backpack condition, speeds with the same letter are not significantly different.
*Signifies differences between the backpack and no backpack condition at a specific walking speed.

Net and peak counterclockwise lower body torque is the sum of all the lower body torque exerted in the clockwise direction. As expected, increasing walking speed was associated with an increase in net (Table 15) and peak (Table 16) counterclockwise lower body torque. In addition, counterclockwise lower body torque was less in the backpack condition than in the no backpack condition. Results for net and peak clockwise torque were equivalent to those for counterclockwise torque, but are omitted to prevent repetition.

Table 15. Net counterclockwise lower body torque, $T_{LBnetCCW}$ (N·m): mean (standard error).

	0.6 m·s^{-1}	**0.8 m·s^{-1}**	**1.0 m·s^{-1}**	**1.2 m·s^{-1}**	**1.4 m·s^{-1}**	**1.6 m·s^{-1}**
Backpack	69.50* (2.57) F	95.32* (2.69) E	119.01* (3.49) D	136.92* (3.12) C	151.17* (2.62) B	223.50* (4.82) A
No Backpack	129.49 (2.50) F	157.15 (2.26) E	181.71 (2.28) D	206.11 (2.75) C	245.44 (3.36) B	298.63 (5.95) A

Main Effect of Backpack: 0.0001
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: 0.3300
Within a backpack condition, speeds with the same letter are not significantly different.
*Signifies differences between the backpack and no backpack condition at a specific walking speed.

Table 16. Peak counterclockwise lower body torque, $T_{LBPeakCCW}$ (N·m): mean (standard error).

	0.6 m·s^{-1}	**0.8 m·s^{-1}**	**1.0 m·s^{-1}**	**1.2 m·s^{-1}**	**1.4 m·s^{-1}**	**1.6 m·s^{-1}**
Backpack	4.26* (0.18) F	5.93* (0.18) E	7.46* (0.26) D	8.27* (0.21) C	9.37 (0.17) B	12.94 (0.27) A
No Backpack	7.63 (0.19) F	9.20 (0.17) E	9.89 (0.14) D	10.42 (0.15) C	12.08 (0.19) B	15.87 (0.36) A

Main Effect of Backpack: 0.0013
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: 0.8365
Within a backpack condition, speeds with the same letter are not significantly different.
*Signifies differences between the backpack and no backpack condition at a specific walking speed.

As expected from the decrease in thoracic rotation observed in the backpack condition, counterclockwise thoracic acceleration was less in the backpack condition than in the no backpack condition (Table 17). Results for clockwise acceleration were equivalent to those for counterclockwise, but are omitted to prevent repetition.

Table 17. Counterclockwise thoracic angular acceleration, $\ddot{\phi}_T$ (rad·s^{-1}): mean (standard error).

	0.6 m·s^{-1}	**0.8 m·s^{-1}**	**1.0 m·s^{-1}**	**1.2 m·s^{-1}**	**1.4 m·s^{-1}**	**1.6 m·s^{-1}**
Backpack	1.93* (0.10) F	2.56* (0.04) D	3.52* (0.06) D	4.17* (0.06) C	5.04* (0.08) B	6.26* (0.12) A
No Backpack	4.40 (0.08) D	5.77 (0.08) C	7.26 (0.10) B	8.13 (0.10) A	8.23 (0.10) A	8.34 (0.15) A

Main Effect of Backpack: <0.0001
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: 0.0016
Within a backpack condition, speeds with the same letter are not significantly different.
*Signifies differences between the backpack and no backpack condition at a specific walking speed.

Net and peak counterclockwise, positive upper body torque is the sum of all the upper body torque exerted in the clockwise direction. As expected, increasing walking speed was associated with an increase in net (Table 18) and peak (Table 19) counterclockwise upper body torque. Because adding the backpack increased the MOI of the upper body, an increase in upper body torque was observed despite the decrease in thoracic angular acceleration. Results for net and peak clockwise torque were equivalent to those for counterclockwise torque, but are omitted to prevent repetition.

Table 18. Net positive thoracic torque, $T_{UBnetCCW}$ (N·m): mean (standard error).

	0.6 m·s^{-1}	**0.8 m·s^{-1}**	**1.0 m·s^{-1}**	**1.2 m·s^{-1}**	**1.4 m·s^{-1}**	**1.6 m·s^{-1}**
Backpack	133.54* (3.27) F	174.48* (2.88) E	224.64* (3.57) D	261.21* (4.13) C	301.04* (4.94) B	406.40* (7.36) A
No Backpack	68.90 (1.28) F	94.28 (1.23) E	112.21 (1.21) D	124.59 (1.10) C	132.01 (1.21) B	137.54 (1.85) A

Main Effect of Backpack: <0.0001
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: <0.0001
Within a backpack condition, speeds with the same letter are not significantly different.
*Signifies differences between the backpack and no backpack condition at a specific walking speed.

Table 19. Peak positive thoracic torque, $T_{UBPeakCCW}$ (N·m): mean (standard error).

	0.6 m·s^{-1}	**0.8 m·s^{-1}**	**1.0 m·s^{-1}**	**1.2 m·s^{-1}**	**1.4 m·s^{-1}**	**1.6 m·s^{-1}**
Backpack	7.12* (0.22) F	9.58* (0.16) D	13.38* (0.24) E	15.58* (0.24) C	18.85* (0.33) B	23.89* (0.50) A
No Backpack	3.69 (0.08) E	4.93 (0.08) D	6.12 (0.08) C	6.98 (0.09) B	7.18 (0.09) B	7.51 (0.14) A

Main Effect of Backpack: <0.0001
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: <0.0001
Within a backpack condition, speeds with the same letter are not significantly different.
*Signifies differences between the backpack and no backpack condition at a specific walking speed.

The net torque of the body (T_{Net}) represents the sum of all torque acting on the trunk per stride. Increasing walking speed resulted in an increase in net body torque, and as expected, adding a backpack also resulted in an increase in net body torque (Table 20).

Table 20. Net body torque, T_{Net} (N·m): mean (standard error).

	0.6 m·s^{-1}	**0.8 m·s^{-1}**	**1.0 m·s^{-1}**	**1.2 m·s^{-1}**	**1.4 m·s^{-1}**	**1.6 m·s^{-1}**
Backpack	346.50 (7.80) F	450.13 (7.81) E	567.90* (8.97) D	656.58* (10.77) C	766.34* (12.67) B	1166.96* (20.62) A
No Backpack	341.98 (5.65) F	421.98 (5.57) E	474.00 (6.84) D	531.40 (7.95) C	621.81 (9.42) B	733.49 (14.75) A

Main Effect of Backpack: 0.0002
Main Effect of Speed: <0.0001
Interaction of Speed * Backpack: <0.0001
Within a backpack condition, speeds with the same letter are not significantly different.
*Signifies differences between the backpack and no backpack condition at a specific walking speed.

DISCUSSION

In the present study it was hypothesized that compared to unloaded walking, load carriage decreases transverse pelvic and thoracic rotation, mean relative phase between pelvic and thoracic rotations, increases hip excursion, and these changes would be associated with a decreased stride length and increased stride frequency. Our findings supported these hypotheses. In addition, it was hypothesized that carrying a backpack would result in an increase in upper body torque (due to the increased MOI associated with the load), an increase in lower body torque and an increase in net body torque. Our findings supported the hypotheses of an increase in upper body torque, and an increase in net body torque; however, there was a decrease in lower body torque in the backpack condition.

Dimensionless analysis confirmed a lesser contribution of pelvic rotation and a greater contribution of hip excursion to stride length in all velocity conditions while carrying a backpack. However, the increase in hip excursion during load carriage does not fully compensate for the decreased pelvic rotation; an increase in stride frequency is required to maintain a constant walking speed. A significant effect of load carriage on the ratio between thoracic and pelvic rotation ($DThR$) was found, indicating that at all walking speeds except 1.6 $m \cdot s^{-1}$, the relative amplitude of transverse pelvic was greater than the relative amplitude of thoracic rotation. Finally, dimensionless analysis indicated significant differences in the timing of the trunk rotations (DTR) between load carriage and the unloaded walking. It was concluded that the shorter stride length and higher stride frequency observed when carrying a backpack is the result of decreased pelvic rotation.

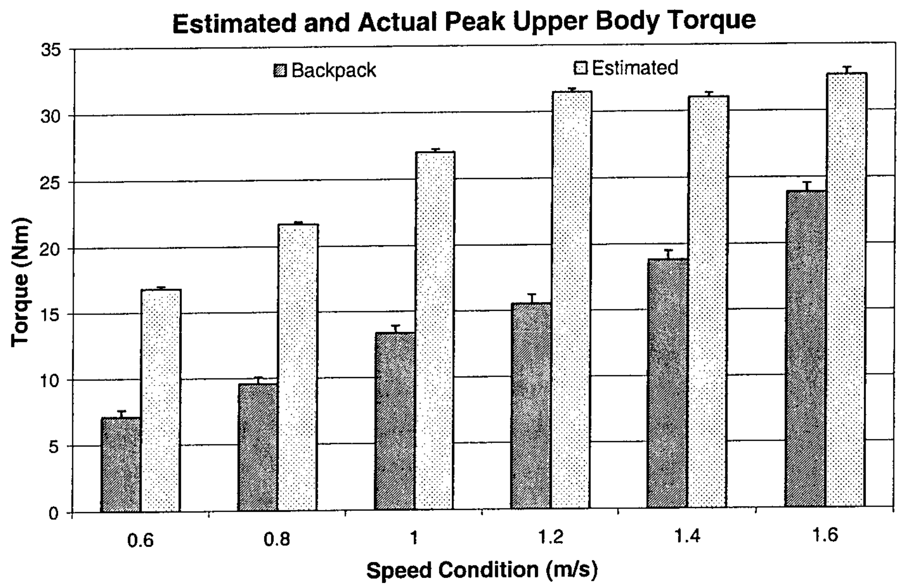
Consistent with the findings of Wagenaar and Beek (19), increasing walking speed during unloaded walking was associated with a linear increase in stride length, stride frequency, and transverse trunk rotation. In addition, this study confirmed that during unloaded walking there is an increase in transverse pelvic rotation observed when walking speed increases greater than 1.0 $m \cdot s^{-1}$ (10), and a gradual transition from a more in-phase pattern of pelvic and thoracic rotation to a more out-of-phase pattern at approximately 0.8 $m \cdot s^{-1}$ (17). Wagenaar and van Emmerik (20) have shown similar changes with increasing walking speed in the relative phase dynamics of inter-limb coordination. In the present study, no transition from a more in-phase pattern to a more out-of-phase pattern of pelvic and thoracic rotation was observed during load carriage.

Previous research suggests the upper and lower torsos counter-rotate at higher walking speeds to reduce the net angular momentum of the body (18). The reduced transverse plane pelvic rotation during load carriage may, in turn, reduce the angular momentum of the lower body. Hence, counter-rotation between the pelvis and thorax as a means of reducing the net angular momentum of the body may not be necessary.

The results of this study showed that increases in *walking speed* were accompanied by increases in upper and lower body torques, as hypothesized. In

addition, *carrying a backpack* also resulted in an increase in upper body torque and net body torque. The increase in upper body torque while carrying a backpack was less than would be expected to result from the large increase in MOI of the upper body caused by the mass of the backpack (Figure 6). This was due to a decrease in thoracic angular acceleration in the backpack condition compared to the no backpack condition. Contrary to our predictions, the addition of a load also resulted in a decrease in lower body torque.

Figure 6. Estimated and actual peak upper body torque



A comparison of the actual peak clockwise upper body torque for the backpack condition and values of upper body torque estimated from the increase in upper body MOI caused by the addition of the backpack for each speed condition.

The results of this study suggest that during load carriage, strategies are used to minimize torque production in the upper body rather than strategies to counterbalance torques. Increases in torque generation about the upper body would require large muscular forces to control the resulting increase in angular momentum, at a potentially higher metabolic cost. Furthermore, high rotational loads may increase the potential for injury (21). The increase in upper body torque observed in the backpack condition may contribute to the increase in low back injuries reported in the literature (8). Future research should investigate this possibility.

One strategy to reduce upper body torque may be to reduce the torque transmitted to the upper body from the lower body. By decreasing torque production in the lower body, the magnitude of torque potentially transmitted to the upper body would be expected to be limited. In this experiment, a decrease in lower body torque resulting from a reduced pelvic rotation during load carriage was observed. Increases in stride length at higher walking speeds are usually achieved by increased pelvic rotation combined with hip excursion (6). Thus, decreased stride length and increased stride

frequency at higher walking speeds during load carriage may emerge as a consequence of the dynamics required to minimize torque production in the upper body.

The more in-phase pattern of pelvic and thoracic rotation during load carriage may illustrate a second method for limiting torque transmission. The dynamic relationship between the upper and lower body during gait may be modeled as two segments connected by a torsional spring. If it is assumed that the pelvis remains fixed and the upper body rotates on it in the transverse plane, the relationship between torque, displacement, and stiffness is described by Equation 6:

$$I\ddot{\theta} = k\theta \tag{6}$$

$I\ddot{\theta}$ represents the torque stored in the spring, k the stiffness of the torsional spring, and θ the amplitude of the angle between the two segments. Increasing θ or k will increase the amount of torque stored in the spring and, consequently, the potential of the spring to generate torque on either segment.

By maintaining a more in-phase relationship between the upper and lower body, the angular displacement, θ, is reduced and, consequently, the torque stored in the spring is also reduced. Thus, less of the torque generated by one segment will be transmitted to the other. In simple terms, the spring is not stretched and, therefore, transmits less energy (assuming a constant torsional stiffness, k). The lack of counter-rotation seen at higher speeds during load carriage supports this method of reducing torque transmission.

According to the torsional spring model, another method to decrease the transmission of torque from the lower body to the upper body is to limit the torsional stiffness, k. Stiffer systems facilitate the transmission of forces between segments through elastic energy storage in the connecting "spring." In physiological systems, stiffness is influenced by co-contraction of the antagonist pair of trunk rotators (the internal and external obliques abdominus). Future research should assess the possibility that torsional stiffness between the upper and lower body is reduced during load carriage.

Previous literature has found differences in stride length between loaded and unloaded walking at higher walking speeds (about 1.78 $m \cdot s^{-1}$; 9), but not at lower speeds (about 1.35 $m \cdot s^{-1}$; 5, 11). However, we did not find a significant speed x backpack interaction effect on stride length. Obusek et al. (12) and Harman et al. (5) studied the effects of load carriage by full-time soldiers trained to march with a 30-inch stride length. In contrast, neither the subjects for Martin and Nelson (9) nor for our study were full-time soldiers. The apparently differing effects of load carriage on stride length may be due to different subject populations tested.

This study was limited in that it was designed to describe changes in transverse plane kinematics associated with carrying a backpack. Consequently, we cannot conclude causal relations exist between the changes we observed. In addition, we did not take into consideration movement in the frontal plane. For example, an increase in frontal plane pelvic tilt will shift the transverse plane axis of rotation of the hip closer to the ground during swing, requiring increase in hip excursion to maintain stride length, as well as an increase in knee flexion and ankle dorsiflexion to prevent floor contact during mid-swing.

The present study demonstrates that the shorter stride length and higher stride frequency observed during load carriage are associated with changes in transverse plane kinematics. In addition, these changes in transverse plane kinematics (decreased pelvic and thoracic rotation) may emerge as a consequence of the dynamics required to minimize torque production in the upper body which, in turn, may reduce the amount of muscle force required to control the load and the potential for injury.

CONCLUSIONS

There is a decreased stride length during load carriage that is due to a lack of increase in pelvic rotation.

There is an increase in hip excursion to assist in maintaining stride length during load carriage. However, this increase is not sufficient to fully compensate for the lack of pelvic rotation, resulting in a decreased stride length and increased stride frequency.

There is an increase in upper body torque that is less than predicted from the increase in upper body MOI caused by adding the backpack, suggesting there are adaptations in the gait during load carriage that minimize upper body torque.

Minimization of upper body torque may occur in order to reduce the potential for injury from high rotational loads and to reduce the amount of muscle force required to control the upper body and backpack segment.

Lower body torque is less during load carriage than during unloaded walking. This decrease in lower body torque reduces the amount of torque potentially transmitted from the lower body to the upper body.

The decreased pelvic rotation likely results from adaptations in the gait that decrease lower body torque. Consequently, the decreased stride length observed during load carriage may result from attempts to minimize upper body transverse plane torque.

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